Parallel MRI with Extended and Averaged Grappa Kernels (PEAK-Grappa): SNR optimized fast dynamic imaging with high acceleration factors

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Introduction: In order to reduce total acquisition time or to increase spatiotemporal resolution in dynamic MRI several techniques such as TSENSE [1], TGrappa [2], kt-SENSE, kt-BLAST [3] and kt-Grappa [4] have recently been introduced. The combination of parallel MRI and temporal imaging acceleration permits the use of higher acceleration factors compared to conventional parallel imaging techniques, e.g. SENSE and Grappa. Nevertheless, all techniques suffer from decreased SNR and/or increased temporal blurring with higher acceleration factors. Here we present a novel reconstruction for dynamic MRI which optimally utilizes information along all available spatial directions as well as the temporal dimension. A method called PEAK-Grappa (Parallel MRI with Extended and Averaged Grappa Kernels) is presented which is based on an extended spatio-temporal Grappa kernel in combination with temporal averaging. By including Grappa weight estimation along the time dimension while at the same time exploiting the averaging properties of uncorrelated noise of different acquired time frames, it is possible to maintain or even increase SNR while minimizing temporal blurring. Phantom measurements with reduction factors up to 12 and results from volunteer experiments demonstrate that PEAK-Grappa preserves the dynamics of time-resolved measurements while considerably improving SNR compared to conventional methods.

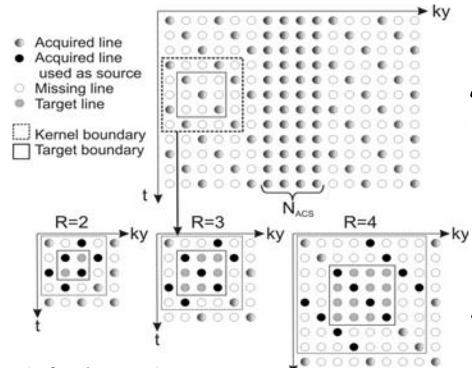
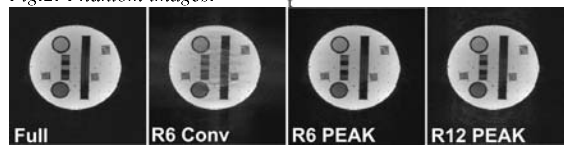


Fig.1: Upper image: acquisition scheme for R=3. Lower images: kernels for different R-factors with source and target lines.

Fig.2: Phantom images.



Methods: For PEAK-Grappa data was acquired using a pre-defined number of N_{ACS} reference lines for each time frame and incrementally shifted phase encoding amplitudes for subsequent time frames resulting in a ky-t space sampling pattern as illustrated in Fig.1. For every reduction factor (R) uniform 3D kernels (two encoding kx, ky directions and one temporal dimension) were defined as exemplary shown in Fig.1 for R=2, 3, and 4. Such a uniform kernel was chosen since different kernels for different *k-t*-data points might lead to systematic errors and hence artifacts in image reconstruction. The reconstruction process is represented by

$$S_c(k_y^{tp}) = \sum_{c,bx,sp} \left(S_c(k_y^{sp}) \cdot \left(\frac{1}{N_{CF} - bt + 1} \sum_{N_{CF}} w^{CF}(c,bx,sp,tp) \right) \right)$$

where *sp(ky,t)* and *tp(ky,t)* represent the source and target points, *bx* and *bt* the kernel size in x- and t-direction N_{CF} the number of cardiac frames and c the counts through the number of coils. Note that Grappa weights w^{CF} are independently determined for each time frame and subsequently averaged (inner bracket of equation) to generate a single unique 3D kernel for the reconstruction of the entire k-t-space. For PEAK-Grappa reconstruction the kernel is shifted by an increment R in ky- and t-direction across the k-t-space. Fig.1 shows the kernels used for calculation of coil weights and reconstruction of the missing lines in k-t-space for R=2-4 with a kernel size in kx-direction *bx*=5. The outer kernel boundary in k-t-space was determined by an R dependent number of source lines which were used to estimate missing target lines within an inner target boundary. After reconstruction the reference lines were copied back into the data matrix. For comparison, conventional Grappa reconstruction was performed with a kernel size of *bx*=5 and *by*=4. All reconstructions were performed with 24 reference lines (26 for conv. Grappa with R=6 due to a kernel width *ky*=25). Due to these reference lines the true acceleration factor is smaller than the reduction factor (e.g. for R=8, 320 ky-lines and 24 reference lines the true acceleration factor is 5.25). To account for the spatial dependency of SNR in parallel imaging, SNR was calculated by averaging and subtracting two adjacent time frames (end-diastolic frames for in-vivo data) and by dividing the mean value in a ROI of the averaged image by the standard deviation in the same ROI in the subtracted image.

All measurements were performed on a 3T system (Trio, Siemens Germany). Static Phantom data were acquired using an 8 channel head coil and a k-space segmented Cine gradient echo sequence with 16 time frames and a full data matrix of 320 x 320. In-vivo measurements in a healthy volunteer were performed using a black blood prepared gradient echo sequence with a temporal resolution of 70 ms acquired during breath-hold. For data acquisition an 8 channel thorax coil and an imaging matrix of 96 x 256 was used. Image reconstruction was performed in Matlab (Mathworks). To evaluate the performance of different reconstruction methods, fully acquired k-space data sets were used to compare the following algorithms: standard (full k-space), conventional Grappa with reduction factors from R2-R6, and PEAK-Grappa data with reduction factors R2-12.

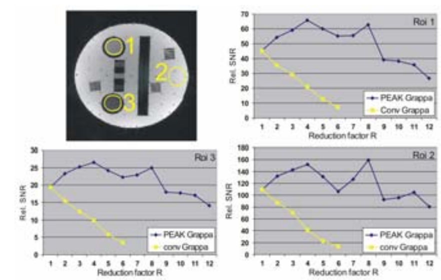


Fig.3: SNR behavior in three different ROI's for conventional Grappa and PEAK-Grappa.

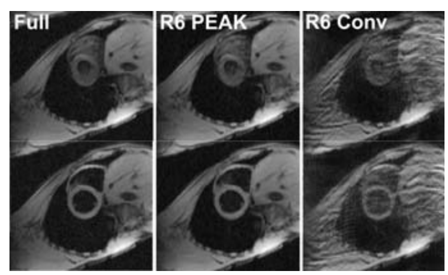


Fig.4: Black blood prepared gradient echo images of the left ventricle; upper row – systole, lower row – diastole.

Results: Fig. 2 shows the phantom images reconstructed with full k-space data, conventional Grappa R=6, and PEAK-Grappa R=6 and R=12. Improved image quality of PEAK-Grappa compared to conv. Grappa can clearly be appreciated. Note that even reconstruction with R=12 provides good image quality with only moderate artifacts. The graphs in Fig. 3 show the relative SNR of the phantom images as a function of reduction factor R in three ROI's shown in magnitude image. As expected, SNR for conv. Grappa reconstructions decreases with increasing reduction factors starting from the full k-space image (R=1). In contrast, SNR for PEAK-Grappa exhibits a completely different behavior. Despite increasing reduction factors SNR can be maintained or even increased until it decreases for highest reduction factors of R=9-12. Results of cardiac Cine images in a normal volunteer are shown in Fig. 4 for a systolic and diastolic cardiac frame reconstructed with full k-space data (a), conv. Grappa R=6 (b), and PEAK-Grappa R=6 (c). The comparison of all methods clearly illustrates the superior image quality and SNR of PEAK-Grappa compared to conv. Grappa. These findings are supported by the relative SNR analysis in Fig.5 demonstrating a similar behavior as in Fig.3. For all regions including the myocardium, SNR can be maintained or even increased for increasing acceleration factors. Generally, SNR decreases for PEAK-Grappa reconstructions when the kernel size reaches the range of the number of time frames. Moreover, by incorporation of temporal weights into PEAK-Grappa, the dynamic information in the Cine images could be preserved as is evident from the absence of temporal burring of the ventricular walls if PEAK-Grappa and full reconstruction are compared in Fig.4.

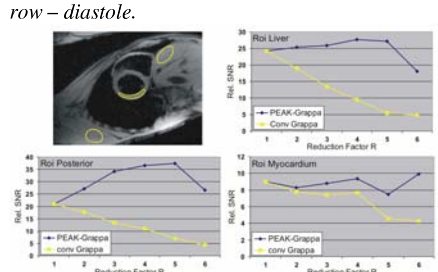


Fig.5: SNR behavior in three different ROI's for conventional Grappa and PEAK-Grappa.

Discussion: The introduced PEAK-Grappa method produces robust image quality to high reduction factors and a higher SNR compared to the full k-space reconstruction. The observed SNR optimization is a result of the averaging process included in the weight estimation used to define a single Grappa kernel for the entire k-t-space. Grappa weight averaging effectively exploits temporally uncorrelated noise in different time-frames and results in considerably optimized SNR performance compared to other parallel imaging techniques while minimizing temporal blurring. The increased number of target points with a decreased noise due to the weight averaging must yield to the slight increase of SNR with higher reduction factors.

References: [1] Kellman et al *MRM* 2001;45:846-52. [2] Breuer et al *MRM* 2005;53:981-85. [3] Tsao *MRM* 2003;50:1031-42. [4] Huang et al *MRM* 2005;54:1172-84.